

Tracking and Monitoring Pulsatility of a Portion of Inferior Vena Cava from Ultrasound Imaging in Long Axis

*Original*

Tracking and Monitoring Pulsatility of a Portion of Inferior Vena Cava from Ultrasound Imaging in Long Axis / Mesin, Luca; Pasquero, Paolo; Roatta, Silvestro. - In: ULTRASOUND IN MEDICINE AND BIOLOGY. - ISSN 0301-5629. - STAMPA. - 45:5(2019), pp. 1338-1343. [10.1016/j.ultrasmedbio.2018.10.024]

*Availability:*

This version is available at: 11583/2731677 since: 2021-09-08T19:15:46Z

*Publisher:*

Elsevier USA

*Published*

DOI:10.1016/j.ultrasmedbio.2018.10.024

*Terms of use:*

This article is made available under terms and conditions as specified in the corresponding bibliographic description in the repository

*Publisher copyright*

Elsevier postprint/Author's Accepted Manuscript

© 2019. This manuscript version is made available under the CC-BY-NC-ND 4.0 license  
<http://creativecommons.org/licenses/by-nc-nd/4.0/>. The final authenticated version is available online at:  
<http://dx.doi.org/10.1016/j.ultrasmedbio.2018.10.024>

(Article begins on next page)

# Tracking and Monitoring Pulsatility of a Portion of Inferior Vena Cava from Ultrasound Imaging in Long Axis

Luca Mesin<sup>a,\*</sup>, Paolo Pasquero<sup>b</sup>, Silvestro Roatta<sup>c</sup>

<sup>a</sup>*Mathematical Biology and Physiology, Department of Electronics and Telecommunications, Politecnico di Torino, Torino, Italy*

<sup>b</sup>*Department of Medical Sciences, University of Torino, Torino, Italy*

<sup>c</sup>*Integrative Physiology Lab, Department of Neuroscience, University of Torino, Torino, Italy*

---

## Abstract

Pulsatility of the inferior vena cava (IVC) provides information on the volume status in healthy subjects and in many clinical conditions. The ultrasound (US) approach to estimate the caval index (CI) is not standardized, as it is operator-dependent and prone to measurement errors due to different factors, including movements of the IVC and non-uniform IVC pulsatility along its longitudinal axis. We propose and test in healthy subjects an innovative automated approach, which tracks the IVC movements registered in a B-mode US video-clip and estimates the pulsatility of an entire portion of the vein rather than of a single arbitrary section. Large variations of CI estimations were found along the longitudinal axis (in the worst case, CI ranged between 15% and 60%), indicating the importance of investigating a whole portion of the vessel.

---

\*Corresponding Author: Luca Mesin, Dipartimento di Elettronica e Telecomunicazioni, Politecnico di Torino, Corso Duca degli Abruzzi, 24 - 10129 Torino - Italy; Email, luca.mesin@polito.it; Phone, +39 011.090.4085

*Keywords:* Inferior vena cava, Ultrasound, Tracking, Caval index

---

## 1 **Introduction**

2 A non-invasive, widely adopted method to assess the intravascular volume  
3 status is based on the pulsatility of the diameter of the inferior vena cava  
4 (IVC), estimated from ultrasound (US) measurements. It found applications  
5 in both healthy subjects (Pasquero et al. (2015)) and conditions of altered  
6 volemic status in patients (Lichtenstein (2005)). However, standardization of  
7 the measurement technique is still lacking (Wallace et al. (2010)). Different  
8 recommendations have been proposed on where to measure the vein diameter  
9 along a longitudinal section (Wallace et al. (2010), Resnick et al. (2011)).  
10 However, the pulsatility of the IVC along its longitudinal axis may not be  
11 homogeneous (Mesin et al. (2015)). As a result, diagnostic recommendations  
12 are non-uniform (Zhang et al. (2014)).

13 Pulsatility is measured in terms of the caval index (CI), reflecting the vari-  
14 ations of vessel diameter during the respiratory cycle (Blehar et al. (2012)).  
15 However, during respiration, the vessel moves relative to the transducer, in-  
16 ducing an additional source of variability in the assessment of pulsatility. For  
17 instance, the M-mode technique allows to monitor changes in IVC diameter  
18 along a fixed line, which is actually fixed with respect to the probe, but may  
19 fluctuate with respect to the vessel, depending on respiratory movements  
20 (Mesin et al. (2015)). Correspondingly, large movement artefacts are pro-  
21 duced, particularly if the vein has an irregular shape (Lichtenstein (2005))  
22 or if it rotates.

23 In a recent paper (Mesin et al. (2015)), we proposed a method for tracking  
24 IVC movements in long-axis US scans and estimating its diameter in each  
25 frame, along a direction moving together with the vein. This approach has a

26 low computational cost and provides a more reliable estimation of IVC local  
27 pulsatility than the standard method (Mesin et al. (2015)). However, the  
28 pulsatility along a single section may be not representative of the dynamics  
29 of the whole IVC. For example, some parts of the vein can show low pulsations  
30 for being anchored to nearby structures (e.g., the diaphragm or vein inlets)  
31 (Wallace et al. (2010)).

32 The non-homogeneous pulsatility of the vessel and the lack of consensus  
33 on an optimal measuring site (Wallace et al. (2010), Resnick et al. (2011)) are  
34 likely to contribute to the contradicting indications found in the literature  
35 (Weekes et al. (2012)). However, investigations on the pulsatile behavior of  
36 the IVC at different longitudinal sites have seldom being reported and were  
37 never based on simultaneous monitoring of a whole IVC segment because  
38 of lack of the necessary computational tools. Here, we propose a new algo-  
39 rithm that tracks the movements and simultaneously monitors the diameter  
40 of different sections of a whole portion of the IVC.

## 41 **Materials and Methods**

42 An algorithm (implemented in MATLAB R2018a, The Mathworks, Nat-  
43 ick, Massachusetts, USA) was developed to process each frame of an US  
44 B-mode video-clip of a longitudinal view of the IVC. Continuous measure-  
45 ments of the diameters along a whole portion of the IVC were computed after  
46 compensating for possible IVC movements.

47 At the first frame of the clip, the user is asked to provide the following  
48 information used by the software for further processing (Figure 1).

- 49 1. A rectangular portion including a longitudinal view of the IVC is se-

- 50 lected.
- 51 2. On this sub-image, the user selects two reference points, assumed as  
52 anchoring sites for the vein. The two reference points are connected by  
53 a reference segment. The reference point on the left is usually close to  
54 the confluence of the hepatic veins into the IVC, the right one is near  
55 the lower hepatic region (caudate lobe) or at the confluence between  
56 the IVC and the portal vein.
  - 57 3. The user then draws the leftmost and rightmost segments cutting the  
58 IVC transversally, along which the first and last diameter measure-  
59 ments are computed (in Figure 1, they are a few mm proximal to the  
60 confluence of the hepatic vein and close to the lower region of caudate  
61 lobe, respectively). The user is then asked to select two points close to  
62 the borders of the IVC along the leftmost section.

63 The software then draws a number of lines (21 in this paper) uniformly  
64 distributed between the leftmost and rightmost borders set by the user.  
65 Specifically, the lines are at the same distance from each other and their  
66 slopes vary linearly between those of the two lines originally selected by the  
67 user. The vein borders are then identified along each of these lines as de-  
68 scribed in (Mesin et al. (2015)), by detecting sharp changes in the intensity  
69 of the US image (which was first processed by a median filter with a square-  
70 shaped mask of 9x9 pixels, in order to smooth the image while still preserving  
71 edge locations). As more than one point can show a sharp change of intensity  
72 along a given section, the one closest to the borders identified in the previ-  
73 ous line was considered. For the leftmost line, this method cannot apply:  
74 for this reason, the user is asked to select points close to the border in the

75 first frame, as mentioned above; moreover, for the subsequent frames, the  
76 borders identified in the previous frame on the first line were used. Once the  
77 superior and inferior IVC borders have been estimated along all intersecting  
78 lines, their profiles were further adjusted by a longitudinal smoothing, which  
79 compensates for minor estimation errors (e.g., noise in the US image, US  
80 artefacts such as reverberation and shadowing). Specifically, for each line,  
81 the border position was re-calculated as the mean between its original value  
82 and the linear interpolation with its two nearest neighbours.

83 The movements of the vein were tracked assuming that they were smooth  
84 in subsequent frames. Moreover, small linear deformations were considered.  
85 The position of each reference point was automatically re-mapped in subse-  
86 quent frames. An estimation of the displacement exhibited by a reference  
87 point from one frame to the next was obtained from the comparison of image  
88 portions (size 128x128 pixels) centred on the current position of the refer-  
89 ence point in the first frame of the pair. The two portions were aligned in  
90 the 2D Fourier domain, to improve resolution (Mesin et al. (2015)). The  
91 same method as in (Mesin et al. (2015)) was considered, but the image por-  
92 tion to be aligned was decomposed into 5 sub-regions. The translations of  
93 all regions were computed and their mean translation and rotation were es-  
94 timated. Moreover, from the third frame on, three images were considered  
95 (the present one and the two previous frames): the movements from each  
96 pair of images were computed imposing that the displacement between the  
97 first and the last was the sum of the two displacements from the first to  
98 the second and from the second to the third. This procedure was found to  
99 provide smoother and more stable movements tracking than using only pairs

100 of subsequent frames.

101     Given the new positions of the reference points, a new reference segment  
102 was calculated, which could appear translated, rotated or stretched compared  
103 to the previous frame. On this basis, each line along which to estimate the  
104 IVC section was re-calculated by keeping constant the angle of intersection  
105 as well as the ratio of the distances between the intersection point and the  
106 two reference points. In this way, these lines followed movements and defor-  
107 mations of the IVC and, ideally, always intersected the IVC along the same  
108 cross-sections.

109     The superior and inferior borders of the IVC along each line and for each  
110 frame obtained as detailed above were affected by high frequency and quan-  
111 tization noises. To improve them, the time series (representing the position  
112 of each point of the borders over time) were low pass filtered with a cut-off  
113 frequency of 4 Hz (this filter and the ones mentioned below were of Butter-  
114 worth type, order 4 and used twice, once with time reversed, to remove phase  
115 distortion and delay).

116     Then, the vessel midline was computed as the average between the two  
117 borders. It was then approximated by a polynomial function of order 4  
118 (hereinafter, the term IVC midline indicates this polynomial approximation).  
119 Then, 10 points were uniformly distributed along the midline of the vein,  
120 excluding the first and last 5% of the curve, to avoid possible edge effects.  
121 Then, the sections orthogonal to the IVC midline passing from each such  
122 points were considered. The IVC diameters in these sections were computed  
123 by interpolation from the estimated vein borders (sampled on the 21 sections  
124 considered by the software). Specifically, the two samples of the considered



125 border closest to the line orthogonal to the IVC midline were identified; then,  
126 the line passing through these two samples was computed; the intersection  
127 point between such a line and the one orthogonal to the IVC midline was  
128 then found. Notice that, in this way, the direction along which to compute  
129 the IVC sections was standardized as suggested in (Pasquero et al. (2015)).  
130 These ten diameters were further considered in the following.

131 In each IVC section, pulsatility was quantified by the CI, defined as the ra-  
132 tio between the range of the estimated diameter time series and its maximum  
133 (Mesin et al. (2015)). CI values were computed from each respiration cycle  
134 (identified based on the low frequency oscillations of the diameter time series  
135 appearing after low pass filtering at 0.4 Hz) and their values were averaged  
136 obtaining a single index of pulsatility for each section. From these estimates,  
137 different indexes could be computed to characterize the overall pulsatility:  
138 the mean pulsatility defined as the mean of the CIs in the different longi-  
139 tudinal sections; the maximum CI, which could indicate possible collapse of  
140 the vein; the standard deviation of the CIs, indicating the variability of the  
141 dynamics along the longitudinal axis.

142 The method was tested on the same data as (Mesin et al. (2015)) (to which  
143 the reader is invited to refer for the details), i.e., on four healthy subjects  
144 in supine position during quiet normal breathing (investigated following the  
145 tenets of the Declaration of Helsinki). Additional tests are shown in the  
146 Supplementary Material. Pulsatility was measured in terms of the CI of  
147 each of the 10 sections orthogonal to the longitudinal axis mentioned above.  
148 The distribution of these CI estimates was then shown in boxplots, showing  
149 median, quartiles and range.

## 150 **Results**

151       Figure 1 shows the data provided by the user (location of the vein, refer-  
152       ence points and left/right range of interest) and the procedure employed by  
153       the algorithm to process each frame.

154       Figure 2 shows an example of processing of data from a subject with  
155       an IVC with an irregular profile. The displacement of the vein and the  
156       time series of the diameters (measured at different locations along the IVC  
157       axis) are shown in 2A and 2B, respectively. Notice the large variability of  
158       section sizes along the axis of the vein. Two frames corresponding to local  
159       minimum and maximum of average section of the vein are shown in 2C and  
160       2D, respectively.

161       Figure 3 shows the results of the processing of the video-clips from the  
162       four subjects. The pulsatility was estimated in terms of CI, which was com-  
163       puted for each of the 10 sections considered. The boxplot in 3A shows, for  
164       each of the four subjects, the distribution of the CI values. Notice that the  
165       distributions are very different across subjects. Specifically, subject 4 shows  
166       the minimum variations of CI along the axis, with a range of about 19-26%;  
167       on the other hand, subject 3 shows the maximum variation of CI along the  
168       axis, with values ranging between 15-60%. Minimum and maximum IVC size  
169       is shown for these two subjects in Figure 3B. Notice that subject 4 has an  
170       IVC with about constant section that indeed pulsates uniformly, whereas the  
171       vein of subject 3 shows low and high pulsatility at proximal (left) and distal  
172       (right) IVC sites, respectively.

173 **Discussion**

174 *The need of exploring an entire portion of IVC*

175 Diameter oscillations of the IVC are not always homogeneously exhibited  
176 in the tract of the vein observed in longitudinal scans, especially in the case  
177 of non-uniform appearance. For example, the retrohepatic IVC may be an-  
178 chored to other structures, like the diaphragm or hepatic vein inlet with a  
179 consequent irregular collapse. Two subjects out of the four considered here  
180 showed large variations of pulsatility in the different sections (subjects 2 and  
181 3, Figure 3). Additional tests are shown in the Supplementary Material,  
182 where other 10 healthy subjects are investigated. Most of them show large  
183 variations of IVC pulsatility in different longitudinal sections.

184 This paper proposes a method to investigate automatically the pulsatility  
185 of the IVC in an entire portion along the longitudinal course of the vein. The  
186 algorithm is an extension of the method proposed in (Mesin et al. (2015)),  
187 which describes the tracking of the vein for the assessment of CI along a sin-  
188 gle IVC section. As compared to the standard CI assessment, based on US  
189 scans in M-mode configuration, IVC tracking proved to considerably reduce  
190 artefacts due to the displacements of the vein in connection with movements  
191 of the diaphragm (Mesin et al. (2015)). In addition to vein tracking, the  
192 novel algorithm allows to compute IVC borders in a region of interest, which  
193 can span several centimetres, depending on the subject's echogenicity. By a  
194 post-processing, it is also possible to estimate the pulsatility along an opti-  
195 mal direction, i.e., orthogonal to the midline of the vessel (Pasquero et al.  
196 (2015)). On the other hand, with the standard M-mode approach, the direc-  
197 tion of the M-line is constrained to originate from the probe and may thus

198 intersect the IVC along a sub-optimal direction. Notice also that, by using  
199 the single diameter studied in (Mesin et al. (2015)), it is not easy for the  
200 operator to select a line orthogonal to the midline, whereas it is simple to  
201 compute it automatically once the IVC borders are available on an entire  
202 portion of the vein along the longitudinal direction. Finally, the possibility  
203 to simultaneously collect and average the CI from the different sections along  
204 the displayed IVC segment reduces the uncertainty related to the arbitrary  
205 choice of a given single section, as done in standard CI assessments.

#### 206 *Perspectives*

207 The proposed algorithm opens new perspectives in the study of IVC pul-  
208 satility. For instance it makes possible to investigate whether the IVC ex-  
209 hibits systematic changes in pulsatility along the longitudinal axis. In ad-  
210 dition, a global CI can be conceived, as the average of the CIs obtained in  
211 the different IVC sections, which may possibly yield a more objective and  
212 repeatable estimation of IVC pulsatility than the standard approach. This  
213 issue is currently under investigation and preliminary results support the  
214 hypothesis (Mesin et al., unpublished observations on 10 healthy subjects  
215 investigated twice by three operators). Further analysis of the multi-section  
216 IVC monitoring may also include the distinct assessment of the respiratory  
217 and cardiac oscillatory components (extracted by filtering the diameter time  
218 series on specific bandwidths). While the latter has already been the object  
219 of some investigation (Folino et al. (2017), Nakamura et al. (2013)), the res-  
220 piratory component has not previously been studied. Moreover, the global  
221 CI was found to be correlated with the right atrial pressure and useful for its  
222 non-invasive estimation, on the contrary of the standard pulsatility estima-

223 tions (Mesin et al., unpublished observations on about 50 patients undergoing  
224 right heart catheterization for measuring the atrial pressure). Finally, recent  
225 works in progress indicate that there is a good correlation of the average CI  
226 (as an overall pulsatility index) with the volemic status of patients (prelimi-  
227 nary study on 64 patients either hypo-, eu- or hyper-volemic).

### 228 *Limitations and possible future improvements*

229 The algorithm is not fully automated, as a few interactions with the user  
230 are required to run the processing. In particular, the small area containing  
231 the vein needs to be indicated. This preliminary step could be removed  
232 by including a method able to identify the IVC automatically (Chen et al.  
233 (2018)).

234 Moreover, the user has to indicate two reference points, which should be  
235 easily tracked. Automated detection of points with maximal discrimination  
236 and invariant under different transformations could be obtained considering  
237 standard image matching techniques, e.g., based on Harris detector, scale  
238 invariant feature transform (SIFT) or speed up robust feature (SURF) (Riha  
239 et al. (2018)). An alternative could be using the popular speckle noise track-  
240 ing to estimate the full motion of the vessel (Krupa et al. (2007)).

241 The present semi-automated implementation opens the problem of assess-  
242 ing the repeatability of the results when the software is run many times, with  
243 different inputs. A general assessment is not possible, as the output depends  
244 on the specific video-clip and on the actual portion of vein which is selected  
245 by the user. However, as a preliminary test, the same video-clip (i.e., the  
246 one recorded from subject 3, showing the largest variations) was processed  
247 8 times, considering different selections of input data. The estimated dis-

248 tributions of CI were very similar. Indeed, the following parameters were  
249 extracted from them (given in terms of mean  $\pm$  standard deviation): mean  
250  $0.416\pm 0.027$ , median  $0.447\pm 0.028$ , standard deviation  $0.168\pm 0.013$ , range  
251  $0.454\pm 0.033$ .

252 A more important limitation is related to the separation between the  
253 US recording and the off-line processing. This introduces a delay in the as-  
254 sessment and also it does not provide immediate feedback about whether  
255 the quality of the US imaging is adequate for the processing. This prob-  
256 lem could be solved by a real time algorithm embedded into the US system,  
257 which would provide the rendering of the estimated IVC borders, guiding  
258 the operators in the acquisition of optimal video-clips. The embedding of  
259 the software in a US machine requires to reduce the computational time.  
260 The present implementation (sequential, interpreted code, implemented in  
261 MATLAB R2018a), when run on an average personal computer (with In-  
262 tel(R) Core(TM) i7-7500U, Double-Core, clock frequency of 2.9 GHz, 8 GB  
263 of RAM and 64 bits operating system), took about 400 ms per frame (i.e.,  
264 about 2 minutes to process a video-clip of 300 frames as those considered  
265 here). The processing time could be largely reduced by considering a com-  
266 piled implementation embedded on a US machine with a powerful processor  
267 (a parallel implementation on a GPU or an FPGA could also be considered).  
268 Notice also that the frequency range of the investigated signal is of a few Hz,  
269 so that the video-clip could be sampled with a frame rate of about 10 Hz,  
270 reducing further the computational cost.

271 **Conclusions**

272 We introduced an algorithm to track the movements and identify the  
273 borders of an entire portion of IVC in long axis US scans. The proposed  
274 method helps to 1) reduce movement artefacts, 2) improve objectivity in the  
275 assessment of IVC pulsatility (as the arbitrary choice of a single IVC section  
276 is eliminated as well as the choice of the respiratory cycle, as different sections  
277 are automatically studied and all breath cycles are identified and the IVC  
278 pulsatility is averaged across them), 3) improve stability and accuracy of the  
279 estimation by measuring perpendicularly to the vessel axis.

280 Simultaneous monitoring of several sections of a whole IVC segment sig-  
281 nificantly extends current US capabilities, providing a longitudinal descrip-  
282 tion of IVC size and pulsatility and revealing possibly relevant regional dif-  
283 ferences in its elastic behavior.

284 An instrument implementing the algorithm described in this paper was  
285 recently patented by Politecnico di Torino and Università di Torino (patent  
286 number 102017000006088).

287 **References**

- 288 Blehar D, Resop D, Chin B, Dayno M, Gaspari R. Inferior vena cava dis-  
289 placement during respirophasic ultrasound imaging. *Critical Ultrasound*  
290 *Journal*, 2012;4:1–5.
- 291 Chen J, Li J, Ding X, Chang C, Wang X, Ta D. Automated identification and  
292 localization of the inferior vena cava using ultrasound: An animal study.  
293 *Ultrason Imaging*, 2018;40:232–244.
- 294 Folino A, Benzo M, Pasquero P, Laguzzi A, Mesin L, Messere A, Porta  
295 M, Roatta S. Vena cava responsiveness to controlled isovolumetric res-  
296 piratory efforts. *Journal of Ultrasound in Medicine*, 2017;36:2113–2123.
- 297 Krupa A, Fichtinger G, Hager G. Full motion tracking in ultrasound using im-  
298 age speckle information and visual servoin. *Proc. ICRA*, 2007:2458–2464.
- 299 Lichtenstein D. *Inferior vena cava. general ultrasound in the critically ill.*  
300 Berlin: Springer, 2005;23:82.
- 301 Mesin L, Pasquero P, Albani S, Porta M, Roatta S. Semi-automated tracking  
302 and continuous monitoring of inferior vena cava diameter in simulated and  
303 experimental ultrasound imaging. *Ultrasound Med Biol*, 2015;41:845–857.
- 304 Nakamura K, Tomida M, Ando T, Sen K, Inokuchi R, Kobayashi E, Naka-  
305 jima S, Sakuma I, Yahagi N. Cardiac variation of inferior vena cava: new  
306 concept in the evaluation of intravascular blood volume. *J Med Ultrasonics*,  
307 2013;40:205–209.



308 Pasquero P, Albani S, Sitia E, Taulaigo A, Borio L, Berchialla P, Castagno F,  
309 Porta M. Inferior vena cava diameters and collapsibility index reveal early  
310 volume depletion in a blood donor model. *Crit Ultrasound J.*, 2015;7:17.

311 Resnick J, Cydulka R, Platz E, Jones R. Ultrasound does not detect early  
312 blood loss in healthy volunteers donating blood. *J Emerg Med.*, 2011;41:270–  
313 275.

314 Riha K, Zukal M, Hlawatsch F. Analysis of carotid artery transverse sections  
315 in long ultrasound video sequences. *Ultrasound Med Biol.*, 2018;44:153–  
316 167.

317 Wallace D, Allison M, Stone M. Inferior vena cava percentage collapse during  
318 respiration is affected by the sampling location: an ultrasound study in  
319 healthy volunteers. *Acad Emerg Med*, 2010;17:96–99.

320 Weekes A, Lewis M, Kahler Z, Stader D, Quirke D, Norton H, Almond C,  
321 Middleton D, Tayal V. The effect of weight-based volume loading on the  
322 inferior vena cava in fasting subjects: a prospective randomized double-  
323 blinded trial. *Acad Emerg Med.*, 2012;19:901–907.

324 Zhang Z, Xu X, Ye S, Xu L. Ultrasonographic measurement of the respiratory  
325 variation in the inferior vena cava diameter is predictive of fluid respon-  
326 siveness in critically ill patients: Systematic review and meta-analysis.  
327 *Ultrasound Med Biol*, 2014;40:845–853.

328 **Figure Captions**

329 **Figure 1:** Example of user setting (A-C) and automated processing (D-E)  
330 on the first frame of a US video-clip showing a longitudinal view of the  
331 IVC. A) Selection of a rectangle including the IVC portion of interest.  
332 B) Enlargement of the selected portion, used for further processing.  
333 C) Reference points (squares), leftmost and rightmost sections of inter-  
334 est (continuous lines) and points close to the vessel edges along the  
335 leftmost section (indicated by circles). Based on these settings, the  
336 program defines the reference segment (dashed line) used to track the  
337 IVC displacement in subsequent frames. Notice that the image was  
338 filtered (by a median filter). D) Automated processing of the algo-  
339 rithm: 21 lines are uniformly distributed between the extreme sections  
340 indicated by the user. Along these lines, the profile of the vein is iden-  
341 tified (the estimated border points are indicated with small circles). E)  
342 Post-processing: from the estimated border of the vessel, the midline  
343 is computed and interpolated with a polynomial function of order 4  
344 (curvilinear line); ten equidistant points are selected on this function  
345 and new lines perpendicular to it are considered as sections along which  
346 the vein diameters are evaluated (border points indicated with small  
347 circles).

348 **Figure 2:** Example of processing of a video-clip. A) Displacement of the  
349 IVC shown in terms of the trajectory of the centroid of the reference  
350 segment. B) Diameters of 10 sections of the IVC orthogonal to the mid-  
351 line of the vessel (grey lines) and mean value (black line). C) Frame of  
352 the video-clip corresponding to a local minimum of the average section.

353 D) Frame of the video-clip corresponding to a local maximum of the  
354 average section.

355 **Figure 3:** Results of the processing of video-clips from four subjects. A)  
356 Distributions of CI along the IVC axis, for each subject (showing me-  
357 dian, quartiles and range). B) Two frames of the video-clips of the  
358 subjects showing minimum and maximum CI variability (subject 4 and  
359 3, respectively).

360 **SUPPLEMENTARY MATERIAL**

361 Results in addition to those shown in the paper are here presented. Ultra-  
362 sound (US) data were recorded from other 10 healthy volunteers (5 females,  
363 5 males; age, mean $\pm$ std 30 $\pm$ 13 years, height 172 $\pm$ 12 cm, weight 63 $\pm$ 11 kg)  
364 with a SonoSite M-Turbo system (SonoSite, Bothell, USA; frame rate 30 Hz,  
365 resolution about 0.4 mm per pixel, 256 grey levels) equipped with a convex  
366 2-5 MHz probe. Two-dimensional (B-mode) longitudinal views of the inferior  
367 vena cava (IVC) were taken with a subxifoideal approach, with the subject in  
368 the supine position during relaxed normal breathing. All subjects provided  
369 written informed consent for the collection of data and subsequent analysis,  
370 according to the Declaration of Helsinki. The experiment was conducted  
371 within a study on the repeatability of IVC pulsatility estimation (mentioned  
372 in the section Perspectives within the Discussion of the main part of the  
373 paper). Different operators repeated twice the acquisition of US video-clips,  
374 but only single measurements from a single operator are here considered.  
375 The distributions of the caval index (CI) measured along different sections  
376 (orthogonal to the estimated axis of the IVC) are shown in Figure I, for each  
377 subject. The CI distributions indicate that there is a large variability among  
378 subjects: some of them exhibit little variability of pulsatility along the vessel  
379 (like subject number 9) and others have large variations (e.g., subjects 5 has  
380 a range of CI of about 25-80% and subject 6 shows a CI range of about  
381 25-95%). These additional results further support the main thesis of the pa-  
382 per: a careful characterization of the dynamics of the IVC requires exploring  
383 an entire portion of the vessel, otherwise the assessment of the patient will  
384 strongly depend on the specific section considered.

385 **Figure Caption**

386 **Figure I:** Distribution of CI (median, quartiles and range) considering dif-  
387 ferent longitudinal sections for 10 healthy subjects.